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Actuator design using Electroactive Polymers

Diego Fernández [a], Luis Moreno [b], Juan Baselga [b]

a EADS Astrium CRISA,C/ Torres Quevedo, 9 (P.T.M.),28760 Tres Cantos – Madrid, Spain
b Universidad Carlos III de Madrid, Avda. de la Universidad, 30,29811 Leganés – Madrid, Spain

ABSTRACT

In order to make EAP actuators technology scalable a design methodology for polymer actuators is required. Design variables, optimization formulas and a general architecture are required as it is usual in electromagnetic or hydraulic actuators design. This will allow the development of large EAP actuators from micro-actuator units, specifically designed for a particular application. It will also help to enhance the EAP material final performance. This approach is not new, since it is found in Nature. Skeletal muscle architecture has a profound influence on muscle force-generating properties and functionality. Based on existing literature on skeletal muscle biomechanics, the Nature design philosophy is inferred. Formulas and curves employed by Nature in the design of muscles are presented. Design units such as fiber, tendon, aponeurosis, and motor units are compared with the equivalent design units to be taken into account in the design of EAP actuators. Finally a complete design methodology for the design of actuators based on multiple EAP fiber/sheets is proposed. In addition, the procedure gives an idea of the required parameters that must be clearly modeled and characterized at EAP material level prior to attempt the design of complex Electromechanical Systems based on Electroactive Polymers.

Keywords: EAP, actuator, EAP actuator design, biomimetic, artificial muscles.

INTRODUCTION

Emerging EAP actuators, so called artificial muscles [1], are unique compare to traditional actuation technologies. Some of their most important differences are: a distributed actuation, noiseless, flexible, customized deformation, and in some cases built-in sensing and therefore intelligence [2]. As in muscular tissue an electrochemical process is known to be behind the behavior of these particular materials. But although significant steps have been taken at the chemical level, and certain EAPs are today comparable to muscular fibers, a topic still to be addressed is the scalability of the technology.

If EAP actuators are to be applied to such an enormous range of applications as smart structures, bio-inspired robotics, or biomedical devices, a general approach for the design of each type of mechanism is desirable. A small set of general design rules are responsible for the optimization of each design in many familiar fields such as microelectronics, hydraulic systems, electromagnetic actuators...

To approach the scalability of intelligent materials such as EAP actuators in a stepwise manner there are currently two lines of research differing in the dimension of the actuator unit. The first line of research attempts to solve the scalability problem at a material level by looking for new electroactive macromolecules that perform intrinsically as molecular actuators. By assembling several of these molecules an actuator material, with customized behavior and ready for mass production can be engineered.

The other line of research uses existing EAP actuating fibers or sheets, fabricated in the mm or μm range as actuator unit. The actuator unit is assembled with other actuator units as to form a larger/stronger/faster actuator. In previous works, the configuration of the actuator was based on the author's intuition, and in general it was based on the application targeted and on the type of actuation unit rather than in a previously established methodology for the selected actuation unit. Nevertheless, many configurations, based on different actuator units, and seeking different applications have been presented, some of them are:

Configurations based on expanding actuator sheets (i.e dielectric elastomers)

- Roll, Tube, Bow-tie, spider or Diaphragm, configuration, seeking linear actuation and pre-strain of the actuation units [4-6].
- Multiple films based roll, seeking multiple degrees of freedom actuation and pre-strain of the actuation units [7]
- Bimorph or Uniphorm sandwich configuration, seeking bending actuation, and pre-strain of the actuation units[5].
- Multilayer diaphragms stacking seeking antagonistically driven linear actuation [8]

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- Planar array of diaphragms seeking multiple degrees of freedom actuation [9]
- Compliant structure with leaf strings, seeking linear actuator and pre-strain of the actuation unit [10]

Configurations based on bending actuator sheets (i.e. IPCC, CP, CNT)
- Multilayer stacking, seeking alternate actuator and sensing sheets or increased actuation force [11].
- Parallel arrangement, coupling actuator and sensing [12]
- Arch configuration made of two independent bending sheets, seeking an elliptic friction drive (EFD) element [13]
- Coil type configuration seeking linear actuation [14]
- Opposing free-end simultaneously-driven sheets seeking macro/micro manipulation [1,15].
- Opposing simultaneously-driven sheets with both ends attached seeking linear actuation [16]

Configurations based on contractile fibers (i.e. PANi)
- Fiber bundles as to increase force [14]
- Mckibeen configuration exploiting radial expansion of the fibers seeking force amplification [17]

A design methodology for EAP actuators using expanding actuator units was previously reported [16]. The methodology allows the design of the actuator compliant structure. The method considers a compliant mechanism as an assembly of compliant building blocks. Fixed nodes, actuator material force vectors and required degrees of freedom are introduced in the program. Using genetic algorithms, it generates solutions that maximize the highest stroke/stiffness ratio in the output dofs and the highest stiffness in the orthogonal dofs.

In this paper a design methodology valid for contractile fiber units, but extendable to any type of actuation unit is presented. It is derived from the actuator design methodology used by nature to design skeletal muscles. As in many other disciplines, before approaching a design methodology for EAP based actuator and devices a close look at nature design technique might be pretty useful. It is known that muscular fibers differ in terms of velocity and length, but they are similar in terms of relative deformation and force density. It is also known that there are not two identical muscles in nature. This means, that in nature, scalability and adaptability of the actuation system is achieved by playing with the configuration of the elements involved in a muscle (fibers, tendons, and aponeurosis) as well as the control algorithm.

SKELETAL MUSCLE BIOMECHANICS

The muscle is connected in series with tendon(s) and aponeurosis, which attach the entire muscle-tendon unit to the skeletal system. In biomechanics the term “muscle” often refers not only to the muscle fibers but is also used for the entire muscle-tendon unit (MTU) [18]. The filaments of the contractile systems, actin and myosin, reside in a sarcomere overlapping each other. Interaction of these filaments generates force [19]. Sarcomeres are arranged in series to form a single muscle fiber. Several muscle fibers in parallel form fiber bundles or fascicles. Depending on the number of sarcomeres in series within a fiber we talk about long or short fibers [20]. Longer fibers containing more sarcomeres in series tend to have greater range of motion over which they can generate force. In addition, longer fibers can develop faster contraction speed. Arrangement of muscle fibers within a muscle into a parallel alignment or into a certain angle (i.e. pennation) in respect to the muscle’s force-generating axis has been well described in the literature [20-23].

Tension developed by a muscle varies with its length. The isometric force-length relationship obtained during maximal activation at variety of muscle lengths, shows that the greatest forces can be produced at lengths in the middle of the range. Increased compliance within the muscle shifts the relationship to longer lengths [20]. Also, if the force-length relationship is derived using sub maximal activation, the optimum length at which the greatest force can be produced locates at longer muscle lengths [24].

The force that a muscle can produce is dependent on the velocity of its length change. Experimentally, a muscle is stimulated maximally and allowed to shorten (or lengthen) against a constant load. The muscle velocity is measured and plotted against the resistive force as constant value [20]. However, it does not give a realistic measure of the instantaneous power of a muscle, nor work available through a full contraction-relaxation cycle under natural conditions the muscles normally operate [25].

Both the force-length and force-velocity relationships derived using the mentioned experimental approach have very different shapes during natural locomotion. This is because during movements, muscles act with constantly varying length and velocity. Nowadays *in vivo* measurements of such curves are available [18].

Typical values for the Young's modulus of muscle, tendon, and bone are 200Kpa, 1 Gpa, and 20 Gpa. It is clear that bone is the stiffest and muscle the least stiff of the connective tissues [20]. Aponeurosis is less stiff than tendon. By attaching muscles to bones via compliant tendons, the operating range of the MTU is increased. This is because some of the length change that would be required to be taken up by muscle fibers is actually taken up by tendons. In counter, as muscle develops force, tendons will strain allowing the muscle to shorten, reducing control of the movement by the muscle. Tendinous tissues are viscoelastic with hysteresis, creep and force relaxation properties [26]. Besides elastic energy storage, tendons have important effect on muscle fiber length and velocity. Compliant tendons act as a buffer decreasing strain in the muscle and allow muscles to work in high force and low velocity region on the force-velocity relationship [20].

While the functional unit of force generation is the sarcomere, the functional unit of movement is the motor unit. A motor unit is defined as an α-motoneuron plus the muscle fibers it innervates.

A skeletal muscle is characterized by the following properties:

- Force (F)
- Deformation (d)
- Velocity of the deformation (Vd)
- Number of degrees of freedom (n)
- Weight and volume (M, V)
- Muscle compliance.
- Settling time.

Other interesting properties such as work or power are derived from these measurements. Muscle's function within the locomotion system is obtained by optimizing the above properties. Detailed animal studies have contributed to the current knowledge that the muscles are not only motors but also can act as brakes, springs and struts [27].
With the aim of understanding nature design methodology, the mechanical model of muscle proposed by Zajac has been chosen [28]. This mechanical model considers a muscle a collection of equally long fibers in parallel, where all fibers are oriented either in the direction of the tendon (i.e. a parallel-fibered muscle) or at an acute angle $\theta>0$ to the tendon (figure 1). The Single Input Single Output model proposed by Zajac explained the mechanical behavior of muscle as result of a combination of Fiber Length, pennation angle, tendon length, muscle velocity and muscle force. Since muscle velocity and muscle force aren't design variables, we propose to increase the number of parameters involved in the mechanical behavior of muscle to 12 parameters. (Young's modulus of tendon is not taken into account).

- Fiber length (L_F)
- Fiber section (S_F)
- Deformation of the fiber (ΔL_F)
- Fiber contraction velocity (V_F)
- Muscular tissue density (ρ)
- Muscular tissue specific tension(ST)
- Number of fibers per fascicle(N_F)
- Number of fascicles (N_{FC})
- Tendon length (L_T)
- Tendon thickness (S_T)
- Pennation angle (θ)
- Motor Units arrangement.

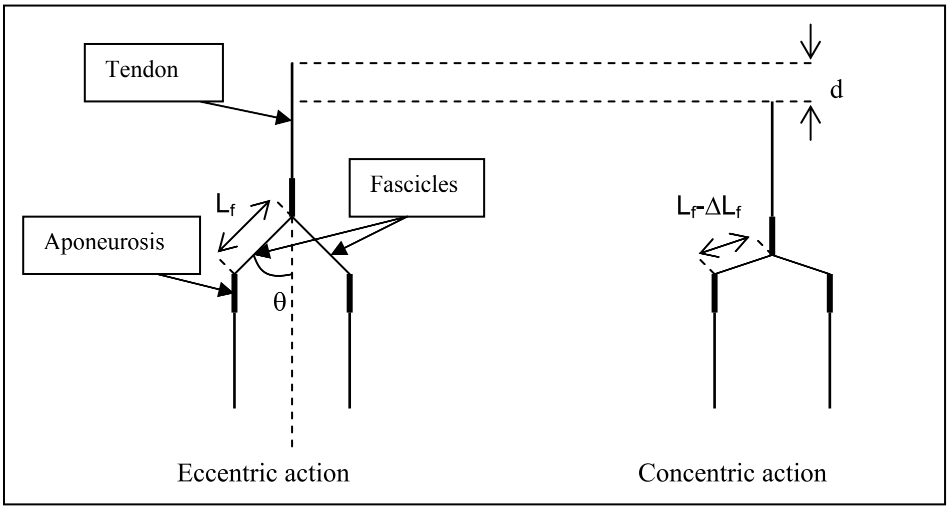

Fig 1. Gans model of skeletal muscle [21].

A commonly reported tension value for mammalian skeletal muscle, when the fiber length is normalized to 2.2 um sarcomere length is 250Kpa. Muscle density is assumed to be in the order of 1.0 g· cm-3 [29]. Contraction velocity of the fibers varies from 17.6um/s in fast fibers to 9.0um/s for low fibers [20].

A widely used relation explaining fascicle force is given by [29]

$$PCSA = \frac{M \cdot \cos(\theta)}{L_f \cdot \rho} \quad [\mathrm{mm}^2] \qquad (1)$$

$$F = PCSA \cdot ST \quad [\mathrm{N}] \qquad (2)$$

where PCSA (Physiological cross-sectional area) is determined from Fascicle mass (M), pennation angle (θ), fiber length L_F, and muscular tissue density ρ. Force developed is then derived using the fibers specific tension ST.
Derived from fig 1 the deformation of the Muscle is governed by equation (3)

$$d^2 - 2 \cdot d \cdot L_f \cdot \cos(\theta) = (\Delta L_f)^2 - 2 \cdot L_f \cdot \Delta L_f; \qquad (3)$$

Discarding the solutions with no physical sense the relation is reduced to equation (4)

$$d = L_F \cdot \cos(\theta) \pm \sqrt{(L_F \cdot \cos(\theta))^2 + (\Delta L_F)^2 - 2 \cdot L_F \cdot \Delta L_F} \qquad (4)$$

$$0 \le \theta < \pi/2$$

Interestingly, from equation (4) we conclude that muscle deformation is governed by three design parameters: angle of pennation, fiber length, and fiber deformation d = d (θ, Lf, ΔLf). As shown in fig 2 the effect of pennation angle is an increase in muscle deformation. With no pennation (θ=0º) deformation would be d= $\cdot \Delta L_f$ The amplification effect of pennation angle has been studied previously [28], but the implications of such parameter in muscle architectural design are much more important than previously assumed. Pennation angle amplifies muscle deformation but also attenuates muscle force by cos(θ) factor. This means there must be an optimum pennation angle, which maximizes deformation while minimizing the loss of force. Such value has been derived from equations (3) and (4) and it is presented in equation 5, and plotted in fig 2:

$$\theta_{dopt} = \arccos\sqrt{2\cdot\left(\frac{\Delta L_F}{L_F}\right) - \left(\frac{\Delta L_F}{L_F}\right)^2} \qquad (5)$$

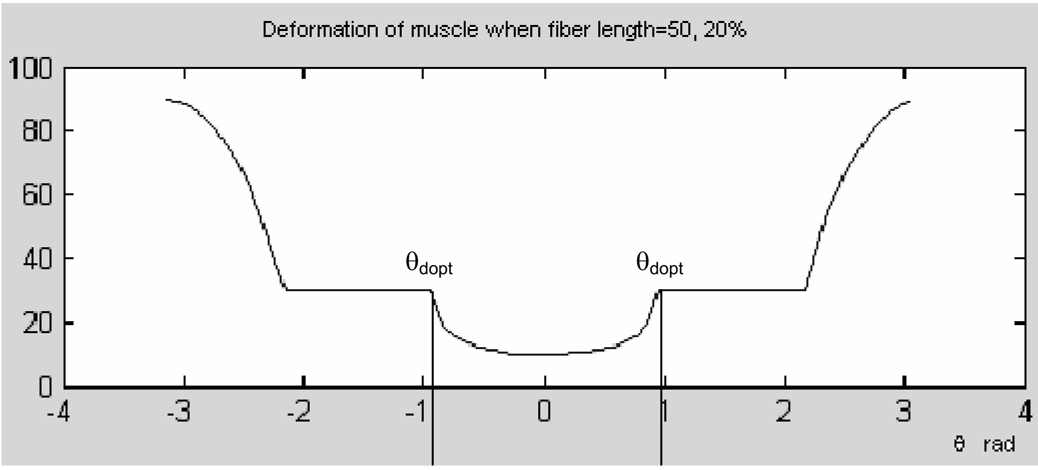

Fig 2. Variation of maximal deformation in muscles with the angle of pennation. For θ=0º deformation is equal to fiber deformation. Above θ=π/2 rad the solution has no physical sense.

By choosing θ_{dopt} the deformation of the muscle is then

$$d\,|\,\theta_{dopt} = L_F\sqrt{2\cdot\left(\frac{\Delta L_F}{L_F}\right) - \left(\frac{\Delta L_F}{L_F}\right)^2} \quad \text{[mm]} \qquad (6)$$

Velocity of the deformation in muscles is governed by the fiber velocity and the pennation angle [23]:

$$V = V_f \cos(\theta) \quad [\mu m/s] \qquad (7)$$

Longer fibers containing more sarcomeres in series tend to have greater range of motion and velocity than shorter fibers. This means that velocity of contraction of muscle is proportional to fiber length.

The number of degrees of freedom of the deformation is determined by the number of angle of pennation involved in the design [21,22].

$$n = \text{Number of different } \theta_i \qquad (8)$$

In muscles with many angles of pennation, the structural design is complex, with fascicles in different planes. During contraction deformation at the interface between fascicle an aponeurosis are produced, inducing complex changes in the shape of the muscle.

When tendons and aponeurosis are neglected the length of the fibers, and the number of fibers define volume (V) and mass (M).

$$V=S_F \cdot L_F \cdot N_F \quad = M/\rho \; ; \qquad (9)$$

where S_F is the fiber cross-sectional area.

Comparing formula (9) with (1) and (7) it is easily inferred that muscle mass or volume are not an indicative of how fast or how much force a muscle can produce. Fiber length and PCSA are the parameters used by physiologist to determine such properties.

Whether a MTU is compliant depends on the ration of its tendon slack length to muscle fiber length [28].

$$J \approx L_T / L_F \qquad (10)$$

Because tendon is elastic, tendon compliance is proportional to tendon slack length. A MTU is to be highly compliant when it has a high tendon slack length to muscle fiber length.

Elastic energy storage in muscle and tendon has been widely analyzed [28]. Elastic energy storage in compliant MTUs (e.g. $L_T / L_F > 10$), resides in tendon and not in muscle. Only for very stiff tendon MTUs will muscle stiffness be a dominant site for storage of elastic energy. Elastic energy storage in tendon may not dominate the energy output of an actuator. Energy output from muscle dominates.
Tendon stiffness can be approximated to [30]:

$$K= S_T \cdot E / L_T \qquad (11)$$

This means that increasing tendon length, the MTU becomes indeed more compliant.

The number of muscle fibers belonging to a motor unit as well as the number of motor units within a whole muscle vary widely [20]. Although muscle fibers belonging to a particular motor unit are scattered over subregions of the muscle, fibers from one motor unit are interspersed among fibers of other motor units. The functional consequence of this dispersion is that the forces generated by a unit will be spread over a larger tissue area. This probably minimizes mechanical stress in focal regions within the muscle. Settling time during muscle activation will depend on the type of control being pursued by the motor units. The nervous systems can vary muscle force output by varying the stimulation frequency to the muscle fibers (called temporal summation in the literature), or by changing the number of motor units that are active at a given time (motor unit recruitment). For relatively low-force contractions, few motor units are activated, while for higher force contractions, more units are activated.
There have been reports of people lifting extraordinary loads under psychological stress. Perhaps this is caused by an extraordinary recruitment of motor units by the central controller, sited in the spinal cord.

ARTIFICIAL MUSCLES DESIGN

In previous section a summary of skeletal muscle biomechanics has been introduced. Design tools employed by nature in order to design the Muscle-Tendon-Unit configuration have been inferred.

But before comparing nature design tools and techniques with those proposed for artificial muscles, considerations about biomimetic engineering must be taken into account. Skeletal muscle configuration is not a closed design. It has been demonstrated that muscle configuration adapts during its life. Length of the fibers or number of fibers in a fascicle can be reconfigured through training [24]. But this is not always desirable. As an example, astronauts under weightlessness for long periods of time exhibit atrophy of the muscular system despite the in-orbit hard physical training.

A conducting polymer fiber differs from a muscular tissue fiber in that neither the velocity of the contraction or the range of motion increase with length. In fact they decrease with length. To solve that problem we will define an artificial sarcomere as a portion of conducting fiber that is attached in series with other sarcomeres to form an artificial fiber (fig 3). By wiring the sarcomeres in parallel, when the muscle fiber in activated all the sarcomeres are activated simultaneously, contracting at the same time, and behaving in a similar way to skeletal muscle fibers. It also allows an easies fiber modeling, since by characterizing the behavior of the artificial sarcomere, the fiber response can be inferred despite the length of the fiber.

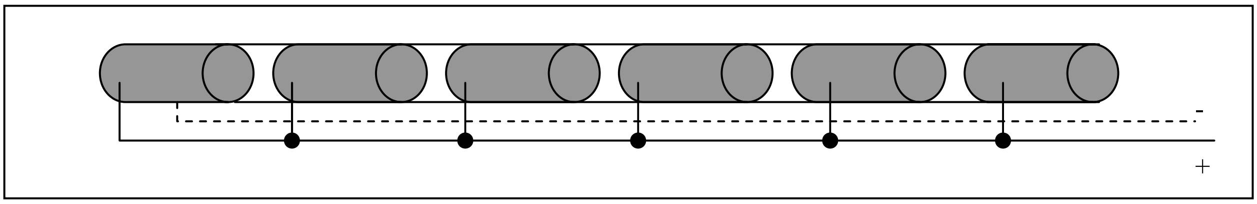

Fig 3. In series connection of contractile fibers of a fixed length, simulating a muscle fiber made of sarcomeres.

Once the artificial fiber has been adapted to its biological counterpart, the skeletal muscle design methodology can be directly employed for artificial muscle design.

The proposed configuration for artificial muscles using contractile fibers is presented in fig 4.

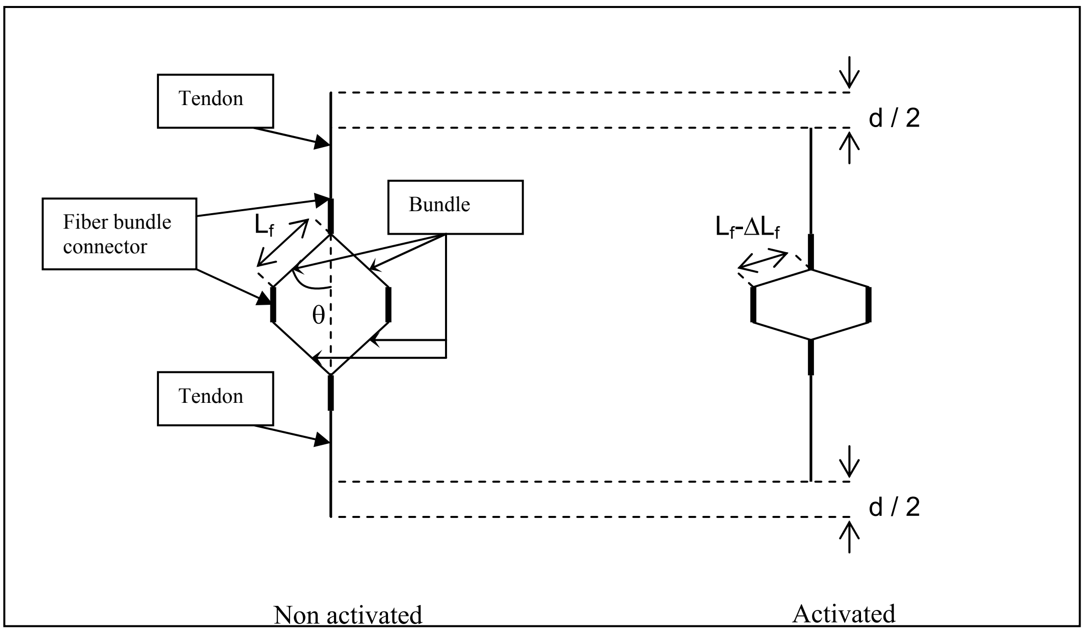

Fig 4. Lineal artificial muscle architecture based on EAP fibers

If more degrees of freedom are required adding more fiber bundles at different angles of pennation can increase the complexity of the configuration.

To explain the methodology the following theoretical problem is presented.

Problem:
Suppose you are asked to design a lineal actuator (1 dof) with a maximum force production of F=5 N, maximum deformation of d= 1cm and rate of contraction of V=0.5 cm/min during maximal force production. The actuator is to be utilized as a motor, and it must be compliant since stretch-shortening cycles (SSC) will be common during the envisioned application.

Approach:
We decide to use PAN fibers. Vf=5%/min, ST= 0.2 Mpa, and deformations of 10 % are easily achievable with such material [1,14]. We will assume the PAN density is 1 g/cm^{-3} and a fiber section of S_f = 1 mm^2
Once the material has been chosen, the design is approached in a stepwise manner.

- Step 1: Selection of the configuration of the actuator.

Since the required actuator is lineal only one angle of pennation is required, and fig 4 is valid as actuator configuration.

- Step 2: Length of the fibers

Due to symmetry of the configuration we require a deformation of 0.5cm. One solution would be to use fibers of 5 cm long and no angle of pennation ($\theta = 0º$). But if we decide to make use of the amplification effects of pennation, we then make use of equation (6) and find out that only L_F = 1.14 cm fibers are required. We must take into account that the proposed configuration involves four fascicles grouped in opposing pairs. This means that every pair of fascicles will be responsible for half the elongation of the actuator.

- Step 3: Angle of pennation.

Using equation (6) implies that the optimum angle of pennation defined by equation (5) has been selected. For fibers with relative deformation of only 10% , equation (5) is applied and the obtained angle of pennation is θ_{opt} = 64º.

- Step 4: Number of fibers

Four fascicles will be responsible for driving the force in the actuator. This means each fascicle will be responsible for 1.25N. Using (1) and (2) we derive that the number of fibers per fascicle is N_F= 29 fibers. A margin of 20% (up to 35 fibers) would be desirable in order to increase the life of the actuator, since the fibers would not require maximal activation in order to derive maximal force. Over dimensioning the number of fibers also allows the configuration of redundant groups of fibers, that by using a "Motor Unit Recruitment" like algorithm (see previous section) would increase the bandwidth of the actuator.

- Step 5: Fiber configuration

In order to obtain a 0.5cm/min velocity, each pair of fascicle must contract at a maximal speed of 0.25cm/min. Using equation (7) We obtain a Velocity of each fiber of V_F=0.57 cm/min
Since fibers are L_F=1.14 cm long, their velocity is only 5%/min = 0.057 cm/min. We require an increase of one order of magnitude in the velocity of the fibers. The solution adopted is to build a fiber made up of 10 fibers of 1.14 mm long connected in series, and driven by the same electrodes (fig 3).

- Step 6: Tendons

In order to assure compliance of the actuator, according to equation (10) two tendons of 10 times the length of the muscle are required.

- Step 7: Mass and Volume

Making use of equation (9), the Mass and volume of the final actuator can be inferred. Using light materials in the tendons, fiber bundles connectors, and taking into account the 10x35x4=1400 wires required to send signals to the 140 fibers predicted the actuator would weight around 10 gr.

CONCLUSIONS

A methodology for the design of artificial muscles using contractile Electroactive Polymer fibers (i.e. Conducting Polymer fibers) has been presented. The methodology is based on a set of EAP material parameters that must be known prior to design: Density of the material, Thickness of the fibers, Specific Tension, Velocity of contraction at maximal activation, and maximal relative deformation. If simulations of the final actuator are to be conducted, accurate understanding and modeling of such parameters, and their variation with driven voltage and environmental variations are necessary. The methodology is only valid for contractile fibers. Future work will include adaptation of the methodology for expanding actuating units, and for bending actuating units.
Application of the methodology for the Computer Aided Design of Artificial Muscles is envisioned in the long term. If EAP actuators are to be utilized by systems engineers to design applications, such a tool is very necessary.
This methodology might also be useful for the mechanical amplification of MEMS actuation.

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